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January 23, 2001



# BOX PCT

Assistant Commissioner for Patents
Washington, D.C. 20231

PCT/DK/99/00034
-filed January 25, 1999

Re: Application of Henning ANDERSEN, Kim Hjortgaard NIELSEN
HEARING AID SYSTEM AND HEARING AID FOR IN-SITU FITTING
Our Ref: Q62611

Dear Sir:

The following documents and fees are submitted herewith in connection with the above application for the purpose of entering the National stage under 35 U.S.C. § 371 and in accordance with Chapter II of the Patent Cooperation Treaty:

- ☑ an executed Declaration and Power of Attorney.
- ☑ a copy of the International Application.
- ☑ 6 sheet(s) of drawings.
- ☐ an English translation of Article 19 claim amendments.
- ☐ an English translation of Article 34 amendments (annexes to the IPER).
- ☑ International Preliminary Examination Report
- ☑ an executed Assignment and PTO 1595 form.
- ☑ a copy of the International Search Report.
- ☑ a Preliminary Amendment

It is assumed that copies of the International Application, the International Search Report, the International Preliminary Examination Report, and any Articles 19 and 34 amendments as required by § 371(c) will be supplied directly by the International Bureau, but if further copies are needed, the undersigned can easily provide them upon request.

Assistant Commissioner of Patents
Washington, D.C. 20231
Page 2
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**PLEASE SEE THE ATTACHED PRELIMINARY AMENDMENT BEFORE CALCULATING THE FEE.**

The Government filing fee is calculated as follows:

| | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|
| Total claims | 16 | - | 20 | = | | x | $18.00 | = | $.00 |
| Independent claims | 2 | - | 3 | = | | x | $80.00 | = | $.00 |
| Base Fee | | | | | | | | | $860.00 |
| **TOTAL FILING FEE** | | | | | | | | | $860.00 |
| **Recordation of Assignment** | | | | | | | | | $ 40.00 |
| **TOTAL FEE** | | | | | | | | | $900.00 |

Checks for the statutory filing fee of $860.00 and Assignment recordation fee of $40.00 are attached. You are also directed and authorized to charge or credit any difference or overpayment to Deposit Account No. 19-4880. The Commissioner is hereby authorized to charge any fees under 37 C.F.R. §§ 1.16, 1.17 and 1.492 which may be required during the entire pendency of the application to Deposit Account No. 19-4880. A duplicate copy of this transmittal letter is attached.

There is no claim to priority.

Respectfully submitted,

David J. Cushing
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**PATENT APPLICATION**

**IN THE UNITED STATES PATENT AND TRADEMARK OFFICE**

| | |
|---|---|
| In re application of | PCT/DK99/00034 |
| Henning ANDERSEN, et al. | Attorney Docket Q62611 |
| Appln. No.: | Group Art Unit: |
| Confirmation No.: | Examiner: |
| Filed: January 23, 2001 | |

For: HEARING AID SYSTEM AND HEARING AID FOR IN-SITU FITTING

**PRELIMINARY AMENDMENT**

Assistant Commissioner for Patents
Washington, D.C. 20231

Sir:

Prior to examination, please amend the above-identified application as follows:

**<u>IN THE SPECIFICATION</u>:**

Page 1, after the title, insert the heading --Background of the Invention--.

Page 5, after line 5, insert the heading --Summary of the Invention--.

Page 7, after line 2, insert the heading --Brief Description of the Drawings--.

Page 8, after line 6, insert the heading --Detailed Description of the Invention--.

**<u>IN THE CLAIMS</u>:**

4. (Amended) Hearing aid system according to claim 1, wherein the hearing aid is a digital hearing aid.

5. (Amended) Hearing aid system according to claim 1, wherein the voltage dividing network comprises at least two fixed value resistors.

6. (Amended) Hearing aid system according to claim 1, wherein the output signal to the receiver is delivered by an digital/analogue converter.

7. (Amended) Hearing aid system according to claim 1, wherein the output signal to the receiver is delivered by a switching amplifier.

8. (Amended) Hearing aid system according to claim 1, wherein the output signal to the receiver is delivered by a bit-stream converter.

9. (Amended) Hearing aid system according to claim 1, wherein the output signal to the receiver is delivered by a $\Sigma$-$\Delta$ converter.

10. (Amended) Hearing aid system according to claim 1, wherein the input signal for the receiver is tapped from the voltage dividing network.

11. Hearing aid according to claim 8, wherein the supply voltage for the amplifier output stage is tapped from the voltage dividing network.

15. (Amended) Hearing aid according to claim 12, wherein the amplifier is a switch mode amplifier and attenuation is achieved by attenuation of the supply voltage for the amplifier output stage.

16. (Amended) Hearing aid according to claim 12, wherein the attenuation is achieved by attenuation output signal from the amplifier.

**IN THE ABSTRACT:**

**Please add the following Abstract of the Disclosure.**

## APPENDIX

**IN THE CLAIMS:**

**The claims are amended as follows:**

4. (Amended) Hearing aid system according to [any one of claims 1, 2 or 3] claim 1, wherein the hearing aid is a digital hearing aid.

5. (Amended) Hearing aid system according to [any one of claims 1 to 4] claim 1, wherein the voltage dividing network comprises at least two fixed value resistors.

6. (Amended) Hearing aid system according to [any one of claims 1 to 5] claim 1, wherein the output signal to the receiver is delivered by an digital/analogue converter.

7. (Amended) Hearing aid system according to [any one of claims 1 to 6] claim 1, wherein the output signal to the receiver is delivered by a switching amplifier.

8. (Amended) Hearing aid system according to[any one of claims 1 to 7] claim 1, wherein the output signal to the receiver is delivered by a bit-stream converter.

9. (Amended) Hearing aid system according to [any one of claims 1 to 8] claim 1, wherein the output signal to the receiver is delivered by a $\Sigma$-$\Delta$ converter.

10. (Amended) Hearing aid system according to [any one of claims 1 to 9] claim 1, wherein the input signal for the receiver is tapped from the voltage dividing network.

11. Hearing aid according to [any one of claims 8 or 9] claim 8, wherein the supply voltage for the amplifier output stage is tapped from the voltage dividing network.

15. (Amended) Hearing aid according to [any one of claims 12 to 14] claim 12, wherein the amplifier is a switch mode amplifier and attenuation is achieved by attenuation of the supply voltage for the amplifier output stage.

16. (Amended) Hearing aid according to [any one of claims 12 to 15] claim 12, wherein the attenuation is achieved by attenuation output signal from the amplifier.

AMENDMENT
Attorney Docket Q62611

**REMARKS**

Entry and consideration of this Amendment is respectfully requested.

Respectfully submitted,

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Date: January 23, 2001

## ABSTRACT

Hearing aid system for the in-situ fitting of hearing aids, said system comprising a separate control device, and a least one hearing aid, adapted for communication with each other, said hearing aid comprising at least one microphone, a signal processor for generating an output signal to a receiver, and means for receiving control signals form the control device. During the in-situ fitting the control device is in communication with said hearing aid for the activation of generation of test signals, which test signals are delivered to said receiver and emitted therefrom as acoustic test signals. Further, the hearing aid comprises a switch means which when said hearing aid is in communication with the control device may optionally be switched between at least a first and a second position, said switch attenuating in the first position the output signal to the receiver using a voltage dividing resistor network, and said switch bypassing in the second position said voltage dividing resistor network so as not to influence the output signal to the receiver.

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Hearing aid system and hearing aid for in-situ fitting.

The present invention relates to a hearing aid system for the in-situ fitting of hearing aids.

For persons with a hearing loss, the sensitivity of the ear will often be frequency dependent within the usual audible range, ie. the person may have almost normal sensitivity at certain frequencies, but a low sensitivity at others.

Since the object of the hearing aid is to give normal hearing at all frequencies, the amplification provided by the hearing aid must as a result also be frequency dependent, with a high amplification at frequencies where hearing sensitivity is low and zero or low amplification where hearing is normal or close to normal.

Because hearing losses vary from person to person the frequency dependency or amplification characteristic for the hearing aid should be adjustable, so that the hearing aid can be fitted to the actual hearing loss of the person.

One way is to separately measure an audiogram for the patient, ie. measuring sensitivity of the ear to different frequencies and sound pressures, using a test signal generator and a headphone, and adjust the settings of the hearing aid accordingly based on the audiogram.

Another way is the in-situ fitting where the audiogram is measured with the hearing aid placed in the ear and acting as an audio signal source instead of the headphone. This is described in eg. US-A-5 710 819.

In the in-situ fitting procedure the hearing aid is coupled to an external control device, with which a generation of test signals for the receiver, ie. the

output transducer of the hearing aid can be activated. The test signals may either be generated in the control device and delivered to the hearing aid, or they may be generated in the hearing aid in accordance with control signals from the control device. In both cases the built-in amplifier of the hearing aid is used to achieve the different levels for the test signals, and hence the output sound levels from the receiver. The control device further may further provide the power for the hearing aid during the fitting procedure.

Even though the use of the hearing aid itself in the fitting procedure has advantages, such as higher accuracy in the fitting of the frequency characteristic compared to the fitting using a separate audiogram, it does have some drawbacks.

One major drawback is that a very high dynamic output range for the acoustic test signals is needed for the fitting procedure.

This dynamic range is expressed as the difference between the maximum output level achievable and the inherent noise level in the amplifier.

The reason that this very high dynamic range is needed is that the amplifier on one hand should be able to deliver signals powerful enough to make the sounds output by the receiver exceed the hearing threshold for persons with severe hearing losses, eg. above 130 dB SPL (Sound Pressure Level). On the other hand, when measuring on persons with normal hearing in at least certain frequency ranges very low sound output levels are needed, and in such cases the inherent amplifier noise should not exceed the level of the test signal. The latter requiring that the amplifier noise does not exceed approximately 10 dB SPL.

Hence, the necessary dynamic range of the amplifier should exceed 120 dB if the hearing aid is to be

fitted in-situ on any person with an unspecified hearing loss.

In fact, if the same amplifier is to be used in different hearing aids of different construction, in particular with different receivers having different responses, the dynamic range should be even higher, eg. 140 dB.

This dynamic range of 140 dB is far more than the dynamic range of 60-80 dB needed under normal circumstances when the hearing aid is used.

Achieving these high dynamic ranges is complex and costly in hardware, and would increase the costs of the amplifier and thus of the hearing aid, whereas lower dynamic ranges of say 90 to 100 dB are readily achieved with both analogue and digital amplifiers. For instance this higher dynamic range would normally in digital hearing aids require a higher number of bits to achieve the higher resolution.

From US-A-3 818 149 and US-A-5 321 758 it is known to attenuate the output signal from the final stage in analogue amplifiers by means of resistor components. However, none of these hearing aids are adapted for in-situ fitting, and hence do not have a need for the mentioned large dynamic range.

In US-A-3 818 149 the attenuation of the analogue signal is done for the purpose of volume control by means of a voltage divider in the form of an adjustable potentiometer. Having such a voltage divider as the final stage before the receiver leads to increased power consumption. Power consumption is an important issue in hearing aids, in particular because these of aesthetic reasons are small, leaving little room for batteries. Having such a voltage divider in the output circuit of a hearing aid is therefore undesirable.

In US-A-5 321 758 is described a programmable analogue hearing aid with multiple frequency bands. When the hearing aid is fitted, the various frequency bands may be attenuated individually. The sum of these individual frequency bands are amplified in an analogue output stage. For the purpose of achieving a desired overall gain of the hearing aid the analogue output signal from the output stage may also be attenuated. This last attenuation is fixed once in the fitting procedure for the hearing aid, and is not changed, unless the hearing aid is fitted anew. This attenuation is achieved by means of a number of resistors which may be connected in parallel with each others between the output of the amplifier and the receiver, ie. in series with the impedance of receiver. The receiver may also be connected directly to the output of the amplifier by short circuiting of all the resistors. Apart from the fact that this way of attenuation also incurs losses, it is further undesirable because the output characteristic of the receiver compared to a solution using a voltage divider will be more dependent on the impedance of the receiver, which may not be linear but depend on frequency.

Contrary to the above mentioned analogue amplifiers digital amplifiers, known as class D or switch mode amplifiers, may, in principle, be made practically loss free. They are therefore often used where there is a need for high efficiency of the amplifier, eg. in battery powered hearing aids. In such amplifiers a fixed voltage level is switched in pulses. The impedance of the receiver receives the full supply voltage during these pulses, giving rise to a current. To achieve a specific output signal the pulses are modulated to give a mean current corresponding to the desired signal. Because the output level may be regu-

lated entirely by adapting the switching cycles there it has never been suggested to use voltage dividers in connection with digital amplifiers as this would compromise the desired high efficiency of the amplifier.

It is an object to provide a hearing aid in which has a dynamic range suited for in-situ fitting, and which overcomes the drawbacks mentioned above.

This object is achieved by splitting the dynamic range of the amplifier into two overlapping reduced ranges, ie. a range for normal use covering eg. from 40 to 130 dB SPL and a low noise range covering eg. from 0 to 90 dB SPL.

In an embodiment according to the invention, this object is achieved with a hearing aid system for the in-situ fitting of hearing aids, said system comprising

a separate control device, and at least one hearing aid, adapted for communication with each other,

said hearing aid comprising at least one microphone, a signal processor for generating an output signal to a receiver, and means for receiving control signals and power from the control device, and

said control device being in communication with said hearing aid during the in-situ fitting for the activation of generation of test signals, which test signals are delivered to said receiver and emitted therefrom as acoustic test signals,

wherein said hearing aid further comprises a switch means which when said hearing aid is in communication with the control device therefrom may optionally be switched between at least a first and a second position, said switch attenuating in the first position the output signal to the receiver using a voltage dividing resistor network, and said switch bypassing in the second position said voltage dividing resistor network

so as not to influence the output signal to the receiver.

The provision of the voltage dividing resistor network allows for operating the hearing aid in two different modes ie. a normal mode and a low noise mode using the one and the same amplifier.

The enlarged dynamic range is then achieved by bypassing the voltage divider in all situations where the enlarged dynamic range is not needed, in particular in normal use of the hearing aid, using only the dynamic range of the amplifier itself, and in situations where the enlarged dynamic range is needed, to use the voltage dividing resistor network to attenuate the output signal from the amplifier, thereby also attenuating the inherent noise of the amplifier.

Since the voltage dividing resistor network is bypassed in all situations except during fitting, the losses incurred by the resistors are of less importance. In particular, they are of absolutely no importance in the case where the control device for the in situ fitting provides the power supply for the hearing aid, which is thus not drawing any power from the limited battery supply.

According to another aspect of the invention the connection between the control box and the hearing aid may, in cases where the control box is not intended to serve as power supply for the hearing aid during the in-situ fitting, take the form of a cordless connection.

A particular aspect of the present invention is the use of a voltage dividing network in connection with a digital amplifier in a hearing aid adapted for in-situ fitting.

The voltage dividing network may according to one embodiment attenuate the output signal from the digital

amplifier, or according to another embodiment, attenuate the supply voltage for the digital amplifier.

The invention will now be described by way of non-limiting examples of embodiments, and in connection with the figures.

In the figures

fig. 1 shows different dynamic ranges,

fig. 2a shows as a diagram an embodiment of the present invention in the normal mode in which the voltage dividing resistor network is bypassed,

fig. 2b shows the same embodiment as in fig 2a, but in the low noise mode in which the voltage dividing resistor network is not bypassed,

fig. 3 shows as a diagram a second embodiment of the present invention,

fig. 4a shows as a diagram a third embodiment of the present invention in the normal mode and with a first polarity of current through the receiver,

fig. 4b shows the third embodiment, but with the opposite polarity of the current through the receiver, compared to fig. 4a,

fig. 4c shows the third embodiment, with the same polarity of the current through the receiver as in fig. 4b, but in a low noise mode,

fig. 4d shows the third embodiment, with the same polarity of the current through the receiver as in fig. 4b, but in a low noise mode,

fig. 4e shows a different way of operating the modulating switches in the third embodiment in the normal mode,

fig. 4f shows a different way of operating the modulating switches in the third embodiment in the normal mode but with the opposite polarity of the current through the receiver compared to fig 4e,

fig. 5 shows an exemplary block diagram of a hearing aid,

fig. 6 shows an exemplary block diagram of a hearing aid with connected control box,

fig. 7 shows another exemplary block diagram of a hearing aid with connected control box.

Fig. 1 shows different dynamic ranges. The column A shows a desired dynamic range of 130 dB SPL. Column B shows typical dynamic range of 100 dB SPL, as can be achieved with most common amplifiers. Column C shows a slightly narrower dynamic range covering the 90 dB from 40 dB to 130 dB. Column D shows another dynamic range of 90 dB, but covering instead from 0 dB to 90 dB, as may be achieved by attenuating the dynamic range of column C by 40 dB. It can be seen that the overlapping dynamic ranges of column C and D will in conjunction provide the desired dynamic range of column A.

Figs. 2a and 2b shows an exemplary embodiment of the present invention. The embodiment incorporates a amplifier, of which only the final stage is shown. In the embodiment shown in figs. 2a and 2b the final stage is a digital/analogue converter 10 of a digital hearing aid, but in principle it could also be the output stage of a fully analogue amplifier or of a switch mode or class D amplifier. To the digital/analog converter 10 is connected a voltage dividing resistor network comprising two resistors 1 and 2, as well as the receiver 5 of the hearing aid. The current through the resistors 1 and 2, is controlled by two switches 3 and 4. Switch 3 being a normally closed switch and switch 4 being a normally open switch. The current flow is indicated with arrows in all of figs. 2a to 4f.

In fig. 2a the normally closed switch 3 short circuits resistor 1 so that the signal from the digital/analogue converter is fed directly to the receiver

5. The normally open switch 4 prevents the resistor 2 from drawing any current from the digital/analogue converter 10. This diagram represents the hearing aid in normal use, ie. the normal mode.

In fig. 2b is shown the diagram representing the hearing aid in the low noise mode, eg. during the in-situ fitting. In this situation the normally closed switch 3 is open and the normally open switch 4 is closed. The current from the digital/analogue converter 10 thus flows though the resistor 1 of the voltage divider and from the tap 21 of the voltage divider partly through the receiver partly through the resistor 2. Hereby the signal to the receiver 5 is attenuated compared with the situation in fig. 2a. Since the signal includes the inherent amplifier noise this noise ia also attenuated.

The current flowing through the resistors 1 and 2 give rise to power loss, but as explained earlier, this is only temporarily during the in-situ fitting, where the power for the hearing aid is often provided by the control box 16. Thus, the power loss is of less or no importance.

Instead of attenuating the output of a digital or class D amplifier as described above, it is in such an amplifier also possible to attenuate the power supply, ie. the supply voltage $U_{CC}$, as will be described in the following.

In fig. 3 is shown an embodiment using a fully digital amplifier of the switch mode type, eg. a class D amplifier. This embodiment is shown in the normal mode only. The use of such a digital amplifier is highly desirable in modern hearing aids because they are generally already digital, ie. using digital signal processing, such as filtering, and because of the high efficiency.

In such a D class amplifier the output current to the receiver 5 is, as mentioned above, not delivered as an analogue signal, but instead as a sequence of high frequency square pulses with alternating positive and negative pulses with a fixed amplitude and a fixed cycle length. The frequency can be several orders of magnitudes higher than the audible frequency which is to be amplified. By regulating the relationship between the width of the positive and negative pulse within the fixed cycle length the mean current in the output signal may be controlled to achieve the desired output signal. This is commonly known as pulse width modulation.

Alternatively the desired output current is achieved by supplying a pulse train of positive or negative pulses of fixed amplitude and length. By variation of the sequence in which the positive or negative pulses appear after each other the mean output current can be regulated. This is commonly known as bit stream modulation.

The embodiment of fig. 3 allows for the use of any of these principles as well as others eg. puls duration/density modulation PDM. The supply voltage $U_{CC}$ in the position shown in fig. 3 is fed through the normally closed switch 3 to the modulating part of the amplifier. The modulating part of the amplifier comprises a first pair of coupled modulating switches 6, 8, a second pair of coupled modulating switches 7, 9 and the receiver 5. The two pairs modulating switches are controlled to give a current of the desired polarity through the receiver 5 in accordance with the above principles. In the situation shown the current will flow from the left to the right through the receiver in the diagram as indicated by arrows. To achieve a current of the opposite polarity the switches 6 and 9

are opened and the switches 7 and 8 closed. It may also be possible to achieve zero current through the receiver 5 by opening all four switches 6 to 9.

In such class D amplifiers it is for a given clock frequency and supply voltage difficult to achieve a low inherent noise because of the discrete square signals with a fixed amplitude is used. To achieve lower noise levels a higher clock frequency or a lower supply voltage must be used.

According to the present invention this low noise mode, which may be necessary in connection with the in-situ fitting of hearing aids with persons having normal hearing in at least some frequency bands, is achieved by attenuating the supply voltage $U_{CC}$.

This is achieved by switching the normally closed switch 3 and the normally open switch 4 to the opposite position of those shown. In this case current will flow through the voltage dividing network comprising the resistors 1 and 2, and the divided supply voltage tapped at the node 21 may be used as supply voltage instead of $U_{CC}$. To achieve the desired output, the modulating switches 6 to 9 must of course be controlled at different switching rates compared with the same signal level in the normal mode, because the reduced supply voltage has to be taken into consideration.

In another embodiment according to fig. 4a to 4f, there may instead of one voltage divider and a one pair of switches 3 and 4 used to bypass it or engage it, respectively, be used two sets of modulating switches. A first set of modulating switches 6 to 9, and a second set of modulating switches 6a to 9a. The first modulating switches 6 to 9 modulate the supply current $U_{CC}$ under normal use in the manner described above. During this, the second modulating switches 6a to 9a may all be open as shown in fig. 4a and 4b, or they may all be

operated in synchronicity with the first modulating switches 6 to 9, as shown in fig 4e and 4f.

In figs. 4a and 4b there is shown one way of operating the modulating switches 6 to 9 in the normal mode. In the normal mode the switches 6a to 9a which are normally open switches are in the open position, allowing no current to flow through the resistors 1a, 1b; 2a, 2b. The modulating switches are operated between the alternate positions shown in figs. 4a, 4b respectively, so as to let current flow through the receiver 5 in alternate directions. If desired, it may also be possible to open all of the modulating switches or at least the modulating switches 6 and 8 to achieve a third state of zero current through the receiver 5.

Referring now to figs. 4c and 4d, when the lower end of the dynamic range, ie. the low noise mode, is needed during the in-situ fitting, the modulating switches 6 and 8 are opened and the modulation of the current is instead effected by means of the modulating switches 6a and 8a in the same manner as described above. The switches 7a and 9a may be closed during this low noise mode or be operated synchronously with the switches 6a and 8a, ie. 7a closing and opening 7a synchronously with 6a and 9a synchronously with 8a, respectively. As indicated by arrows in figs. 4c and 4d current flows in fig. 4c through a first voltage divider comprising the resistors 1b, 2b, and the impedance of the receiver 5. In fig. 4d the modulating switches are in their opposite position compared with fig 4c, and the current flows through a second voltage divider comprising the resistors 1a, 2a and the impedance of the receiver 5. As it can be seen the current flows through the receiver 5 in the opposite direction, ie. gives rise to a pulse of opposite polarity of the one in fig. 4c. In this mode it is of course also

possible to open all of the modulating switches, or at least the modulating switches 6a, 8a or 7a, 9a, respectively, so as to achieve a zero current state.

Figs. 4e and 4f indicate a different way of operating the modulating switches in the normal mode compared to figs. 4a and 4b. Instead of using the switches 6a to 9a as normally open switches, the switches 6a to 9a are moved in phase with the modulating switches 6 to 9. In this case the resistors 1a, 2a; 1b, 2b are either currentless because the switch in series with them is open, or because they are short circuited by the respective modulating switch in parallel with them.

In principle it is also possible with the configuration shown in figs. 4a to 4f to achieve a modulation with 5 levels, ie. full negative, divided negative, zero, divided positive, and full positive, provided that the switches are controlled accordingly.

The switches in all of the embodiments are implemented as electronic switches, eg. semiconductor switches. The control of these switches are known per se, and is merely indicated by the blocks C1a, C2a, C1b, C2b in figs. 4a to 4f.

In a full digital hearing aid the control of the switches may be in accordance with the principles of the amplifier type known as Σ-Δ converter, eg. as the one described in WO-A-96/17493.

In fig. 5 is schematically shown an embodiment digital hearing aid, comprising a pickup or microphone 12 for converting an analogue acoustic signal to an analogue electric signal. The analogue electric signal is digitized in the analogue/digital converter 13 and delivered to a digital signal processor (DSP) 14. From the digital signal processor 14 the signal is delivered to a digital/analogue, which may be a separate element

as described in connection with figs 2a and 2b or it may be the switch mode amplifier itself as described in connection with figs. 3 or 4a to 4f.

Fig. 6 shows schematically an embodiment of a hearing aid adapted for in-situ fitting. For this purpose a control box 16 is connected to the digital signal processor 14 via a control line 17. The control box 16 delivers test signals or controls the generation of test signals, by the digital signal processor 14.

Fig. 7 schematically shows an embodiment of a hearing aid also adapted for in-situ fitting. In this case the control box 16 is connected to the analogue/ digital converter 13 of the hearing aid via a selector switch 20. In the case shown the selector switch 20 is in a position 22 where it delivers the signal from the microphone 12 to the input of the analogue/digital converter 13. If in-situ fitting is desired, the selector switch 20 is moved to eg. the position 19, thereby interrupting the signal from the microphone, and delivering instead the signals from the control box 16 to the analogue/digital converter 13 via the line 17.

In both of the embodiments of figs. 6 and 7 the control box 16 also provide the power for operating the hearing aid during the in-situ fitting.

The control box may eg. be as described in US-A-5 710 819.

If the hearing aid is only to be power supplied via the built-in battery, and not externally from the control box 16, the connection between the control box 16 and the hearing aid may be a cordless connection as indicated by the stapled line 17 in fig. 6, such as an infrared link from the control box 16 to the hearing aid. This is particularly advantageous when the hearing aid itself generates the test signals based on control signals from the control box 16.

Since the enlarged dynamic range A is achieved by two overlapping dynamic ranges C, D each used for a specific situation, it is not necessary to have any adjustment possibility for the attenuation as such. The attenuation can therefore advantageously be achieved with a fixed value only, because this allows for using fixed value resistors 1, 2; 1a, 2a; 1b, 2b, in the voltage dividing network.

## PATENT CLAIMS

1. Hearing aid system for the in-situ fitting of hearing aids, said system comprising

a separate control device, and at least one hearing aid, adapted for communication with each other,

said hearing aid comprising at least one microphone, a signal processor for generating an output signal to a receiver, and means for receiving control signals from the control device, and

said control device being in communication with said hearing aid during the in-situ fitting for the activation of generation of test signals, which test signals are delivered to said receiver and emitted therefrom as acoustic test signals,

wherein said hearing aid further comprises a switch means which when said hearing aid is in communication with the control device may optionally be switched between at least a first and a second position, said switch attenuating in the first position the output signal to the receiver using a voltage dividing resistor network, and said switch bypassing in the second position said voltage dividing resistor network so as not to influence the output signal to the receiver.

2. Hearing aid system according to claim 1, wherein said control device further supplies the power to the hearing aid when said control device being in communication to said hearing aid during the in-situ fitting.

3. Hearing aid system according to claim 1, wherein said control device is in communication with the hearing aid by means of a cordless connection.

4. Hearing aid system according to any one of claims 1, 2 or 3, wherein the hearing aid is a digital hearing aid.

5. Hearing aid system according to any one of claims 1 to 4, wherein the voltage dividing network comprises at least two fixed value resistors.

6. Hearing aid system according to any one of claims 1 to 5, wherein the output signal to the receiver is delivered by an digital/analogue converter.

7. Hearing aid system according to any one of claims 1 to 6, wherein the output signal to the receiver is delivered by a switching amplifier.

8. Hearing aid system according to any one of claims 1 to 7, wherein the output signal to the receiver is delivered by a bit-stream converter.

9. Hearing aid system according to any one of claims 1 to 8, wherein the output signal to the receiver is delivered by a Σ-Δ converter.

10. Hearing aid system according to any one of claims 1 to 9, wherein the input signal for the receiver is tapped from the voltage dividing network.

11. Hearing aid according to any one of claims 8 or 9, wherein the supply voltage for the amplifier output stage is tapped from the voltage dividing network.

12. Hearing aid adapted for in-situ fitting, said hearing aid comprising at least one amplifier,

wherein said hearing aid comprises a first normal mode in which said amplifier operates in a first dynamic range between inherent amplifier noise and maximum output level, and

wherein said hearing aid comprises a second low noise mode, achieved by shifting of the output level range of the first normal mode, thereby attenuating the inherent amplifier noise.

13. Hearing aid according to claim 12, wherein the attenuation is achieved by means of a voltage dividing resistor network.

14. Hearing aid according to claim 13, wherein the resistors of the resistor network have fixed values.

15. Hearing aid according to any one of claims 12 to 14, wherein the amplifier is a switch mode amplifier and attenuation is achieved by attenuation of the supply voltage for the amplifier output stage.

16. Hearing aid according to any one of claims 12 to 15, wherein the attenuation is achieved by attenuation output signal from the amplifier.

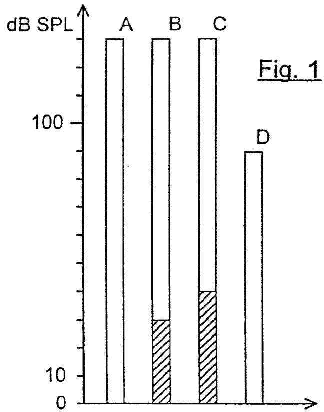



Fig. 1

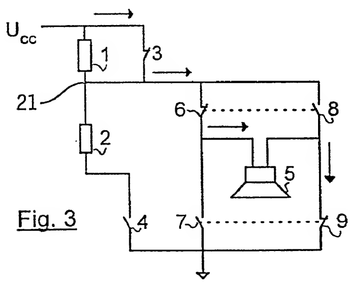



Fig. 3

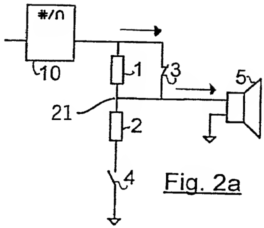



Fig. 2a

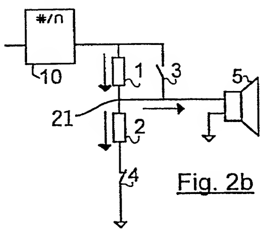



Fig. 2b

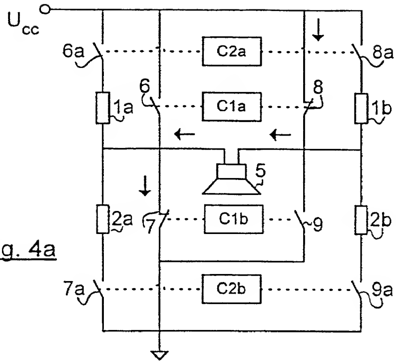

Ucc
6a
C2a
8a
6
1a
C1a
8
1b
5
2a
7
C1b
9
2b
7a
C2b
9a

Fig. 4a

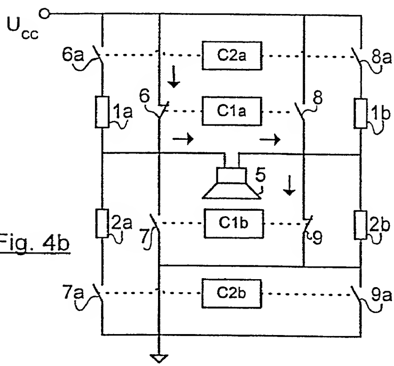

Ucc
6a
C2a
8a
6
1a
C1a
8
1b
5
2a
7
C1b
9
2b
7a
C2b
9a

Fig. 4b

09/744300

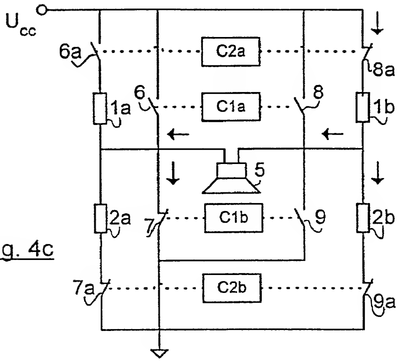



Fig. 4c

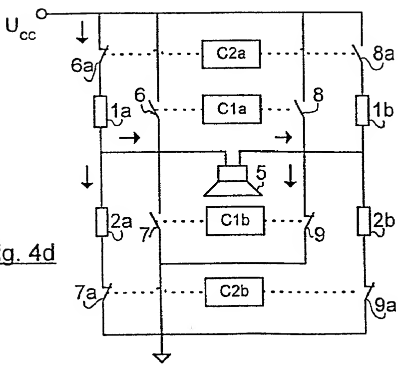
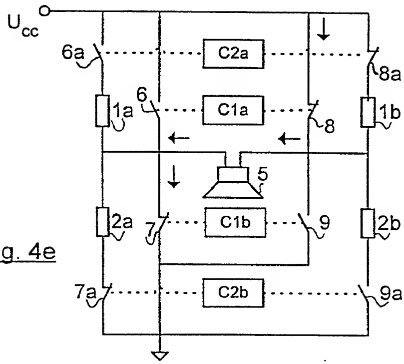



Fig. 4d


$U_{cc}$
6a
C2a
8a
6
1a
C1a
1b
8
5
2a
7
C1b
9
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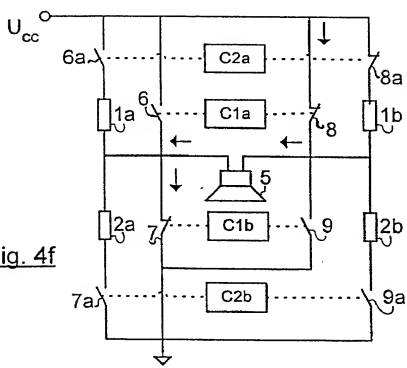
7a
C2b
9a

Fig. 4e


$U_{cc}$
6a
C2a
8a
6
1a
C1a
1b
8
5
2a
7
C1b
9
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7a
C2b
9a

Fig. 4f

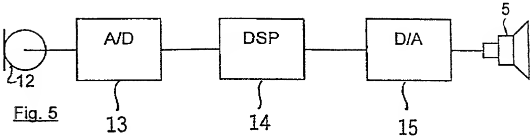


Fig. 5

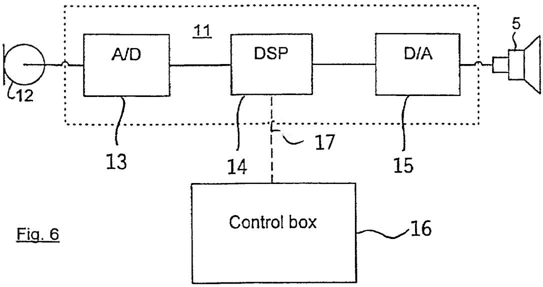


Fig. 6

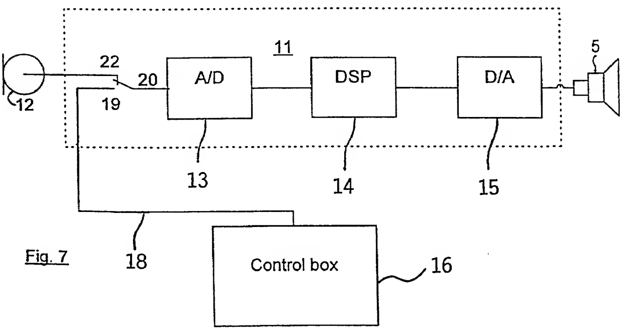


Fig. 7

SOLE/JOINT

# DECLARATION AND POWER OF ATTORNEY

As a below named inventor, I hereby declare that my residence, post office address and citizenship are as stated below next to my name: that I verily believe I am the original, first and sole inventor (if only one name is listed below) or a joint inventor (if plural names are listed below) of the subject matter claimed and for which a patent is sought in the application entitled:

"HEARING AID SYSTEM AND HEARING AID FOR IN-SITU FITTING"

which application is:

☐ the attached application *(for original application)*

☒ application Serial No. PCT/DK99/00034 filed 25 January 1999, and amended on ______ *(for declaration not accompanying application)*

that I have reviewed and understand the contents of the specification of the above-identified application, including the claims, as amended by any amendment referred to above; that I acknowledge my duty to disclose information of which I am aware and which is material to the examination of this application under 37 C.F.R. 1.56; and that I hereby claim foreign priority benefits under Title 35, United States Code §119, §172 or §365 of any foreign application(s) for patent or inventor's certificate listed below and have also identified on said list any foreign application for patent or inventor's certificate on this invention having a filing date before that of the application on which priority is claimed:

| Application Number | Country | Filing Date | Priority Claimed *(yes or no)* |
|---|---|---|---|

I hereby claim the benefit of Title 35, United States Code §120 of any United States application(s) listed below and, insofar as the subject matter of each of the claims of this application is not disclosed in a listed prior United States application in the manner provided by the first paragraph of Title 35, United States Code, §112, I acknowledge my duty to disclose any material information under 37 C.F.R. 1.56 which occurred between the filing date of the prior application and the national or PCT international filing date of this application:

| Application Serial No. | Filing Date | Status *(patented, pending, abandoned)* |
|---|---|---|

I hereby appoint John H. Mion, Reg. No. 18,879; Donald E. Zinn, Reg. No. 19,046; Thomas J. Macpeak, Reg. No. 19,292; Robert J. Seas, Jr., Reg. No. 21,092; Darryl Mexic, Reg. No. 23,063; Robert V. Sloan, Reg. No. 22,775; Peter D. Olexy, Reg. No. 24,513; J. Frank Osha, Reg. No. 24,625; Waddell A. Biggart, Reg. No. 24,861; Robert G. McMorrow, Reg. No. 19,093; Louis Gubinsky, Reg. No. 24,835; Neil B. Siegel, Reg. No. 25,200; David J. Cushing, Reg. No. 28,703; John R. Inge, Reg. No. 26,916; Joseph J. Ruch, Jr., Reg. No. 26,577; Sheldon I. Landsman, Reg. No. 25,430; Richard C. Turner, Reg. No. 29,710; Howard L. Bernstein, Reg. No. 25,665; Alan J. Kasper, Reg. No. 25,426; Kenneth J. Burchfiel, Reg. No. 31,333; Gordon Kit, Reg. No. 30,764; Susan J. Mack, Reg. No. 30,951; Frank L. Bernstein, Reg. No. 31,484; Mark Boland, Reg. No. 32,197; William H. Mandir, Reg. No. 32,156; Scott M. Daniels, Reg. No. 32,562; and Brian W. Hannon, Reg. No. 32,778, my attorneys to prosecute this application and to transact all business in the Patent and Trademark Office connected therewith, and request that all correspondence about the application be addressed to SUGHRUE, MION, ZINN, MACPEAK & SEAS, 2100 Pennsylvania Avenue, N.W., Washington, D.C. 20037.

I hereby declare that all statements made herein of my own knowledge are true and that all statements made on information and belief are believed to be true; and further that these statements were made with the knowledge that willful false statements and the like so made are punishable by fine or imprisonment, or both, under Section 1001 of Title 18 of the United States Code and that such willful false statements may jeopardize the validity of the application or any patent issuing thereon.

Date December 22, 2000 HO

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